Report 1810

AD 608762

DAVID TAYLOR MODEL BASIN
WASHINGTON, D.C.

DEPARTMENT OF THE NAVY

HYDROMECHANICS

O

AERODYNAMICS

O

STRUCTURAL MECHANICS

O

APPLIED MATHEMATICS

O

ACOUSTICS AND VIBRATION

MECHANICAL STIFFNESS OF MAN'S LOWER LIMBS

by

Arthur E. Hirsch
and
Leonard A. White

STRUCTURAL MECHANICS LABORATORY
RESEARCH AND DEVELOPMENT REPORT

October 1964

Report 1810

MECHANICAL STIFFNESS OF MAN'S LOWER LIMBS

by

Arthur E. Hirsch
and
Leonard A. White

October 1964

Report 1810
S-F015 14 04

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

ABSTRACT

Measurements of the compressibility or mechanical stiffness of the lower limbs under static loads are reported. Deformations of the various portions of the limb structure are discussed and major load-bearing regions of the foot are identified.

INTRODUCTION

In the investigation of responses of personnel aboard naval ships during underwater explosion attacks, one area of particular interest has been the effect of high-intensity, short-duration, mechanical shock on the stiff-legged standing man. Knowledge in this area is basic to the establishment of injury mechanisms and maximum tolerance levels for use in predicting human response to various shipboard shock situations and in selecting parameters for the design of protective devices.

Estimates of the natural frequencies associated with the human body have been obtained from vibration experiments[1] and shock tests.[2] In the vibration experiments, stiff-legged standing men were vibrated at various frequencies and their responses observed and measured. In the shock tests, a shipboard shock simulator, developed at the David Taylor Model Basin, was used to simulate deck movements caused by underwater explosions, and the response of the volunteers to such motions was recorded and analyzed. However, since spring constants or mechanical stiffness of the legs could not be established explicitly from either test series, two additional experiments were performed to determine directly the spring constants associated with the lower limbs. This report presents and discusses the results of these tests.

TEST PROCEDURE

The first series of tests was designed to measure the gross movement of the hip bone relative to the floor for stiff-legged men under vertical loadings. Measurements were taken from three male subjects, and the average spring constant of the entire lower limb was computed. The second experiment was performed to determine the stiffness of the foot separately. Here five subjects were utilized, and the displacement of the ankle relative to the floor was measured with a series of loads on the knee. The age, weight, and height of the subjects used in the two experiments are given in Tables 1 and 2.

LOWER LIMB COMPRESSIBILITY

The test consisted of applying various weights to three subjects standing stiff-legged, and measuring the deflection of the top of the hip (iliac crest) relative to the floor (see Figure 1).

---

[1]References are listed on page 17.

TABLE 1

Age, Height, and Weight of Subjects Used in Lower Limb Compressibility Test

Subject No.	Age years	Height inches	Weight pounds
1	29	72	175
2	29	68	170
3	25	66	155

TABLE 2

Age, Height, and Weight of Subjects Used in Foot Compressibility Test

Subject No.	Age years	Height inches	Weight pounds
4	20	77	217
5	23	71	185
6	26	72	185
7	43	70	179
8	19	67	132

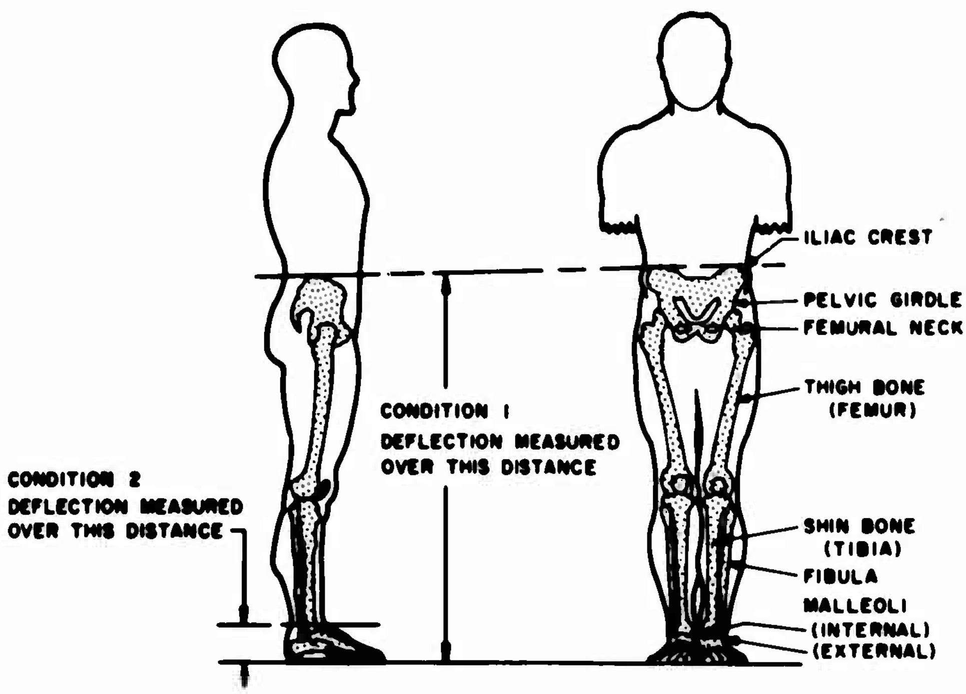

Figure 1 – The Lower Human Skeleton

The subject was taped around the thighs and buttocks to an adjoining support structure which restrained all but vertical movements (see Figures 2 and 3). A belt was drawn about the waist to stretch the skin tautly over the iliac crest and minimize movement of the skin. Cross lines were drawn on a piece of tape which was placed over the iliac crest.

Figure 2 – Experimental Setup for Lower Limb Compressibility Tests

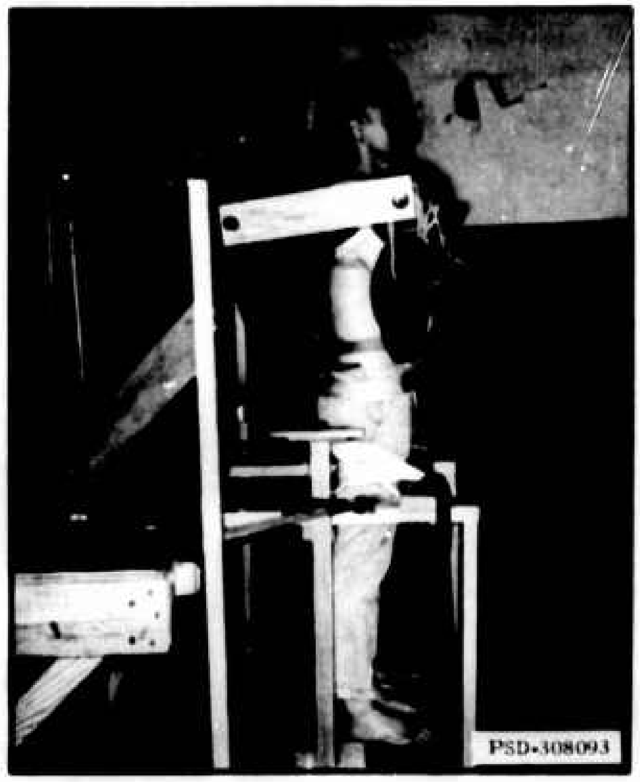

Figure 3 – Side View of Subject during Lower Limb Compressibility Tests

The weights varied from 0 to 280 lb above the subject's body weight. Each test weight was evenly distributed on two bar bells hung from a Dillon weight indicator (Model S with a range of 0–250 lb, in 1/2-lb graduations and a 15-in.-diameter dial).

The bar bells were cross-tied with ropes and attached to a chain-fall which suspended the bar bells over football shoulder pads worn by the subjects; see Figure 2. The load was applied by releasing the tension in the chain-fall in steps so that the weight was gradually taken up by the shoulder pads.

The applied load for each step was determined by taking the difference between each scale reading and the initial reading. It should be noted that the applied load was in addition to normal body weight which was also supported by the lower extremities.

Measurements of iliac crest deflection were made with two height gages, one on each side of the subject. Each pair of deflection readings was averaged to obtain the actual deflection for each load increment. The load was removed and a zero-load reading was taken prior to each new application of the weight.

During each reading (at no load and load conditions), the subject exhaled completely and held as still as possible. Each time the weights were withdrawn, the subject stretched in an effort to return to the initial zero position.

Subjects had been advised to keep their knees stiff but this proved unnecessary since it was found that the natural response to the application of the weight to the shoulders was to support it by stiffening the knees.

FOOT COMPRESSIBILITY

To investigate the response of the foot to vertical loading, three measurements were recorded from five subjects: the total load and the vertical deflections of the inner and outer ankle (internal and external malleoli).

A load table with a pressure transducer hookup and two displacement gages were used in conjunction with related electrical recording equipment to obtain all measurements (see Figure 4). The load table consisted of a sandwich of two aluminum plates 1/4-in. thick with the edges welded all round. The 1/4-in. space between the plates was completely filled with water, and the change in pressure when the plates were loaded was sensed by an appropriate pressure transducer. All loads were applied through a 12- × 12- × 1-in. aluminum plate to maintain a constant-bearing area whether calibrating with lead weights or recording the load on the human foot.

The displacement gages were linear potentiometers; their slides were rigidly attached to each side of a molded fiberglass anklet which was clamped around the subject's ankle. A displacement relative to the ground was recorded from both sides of the ankle so that motions due to ankle rotation could be observed and corrected for. All readings were recorded on an oscillograph which permitted simultaneous recording of deflections and the corresponding loads.

For each run, the subject was positioned in the compression stand (Figure 5) and the leg supported. With the anklet in place and the displacement gages positioned perpendicular to the table, the foot was carefully lowered to contact with the load table. The load was applied parallel to the lower leg through a fiberglass, foam-rubber lined knee cover, and was exerted on the cover by a screw loader. Each subject turned down the screw at his own pace and applied the load until he felt uncomfortable enough to discontinue. (Consequently, the rate of application and the ultimate load vary in each case.) In addition, each subject was advised to relax the leg and make an effort to keep it perpendicular to the table.

Footprints of all subjects were taken at normal stance (with one-half the body weight on each foot), with all of the body weight on one foot, and under the maximum load applied in the compression stand so that changes in the load bearing surfaces of the foot could be observed. Figure 6 shows one subject's footprint for three different knee loads.

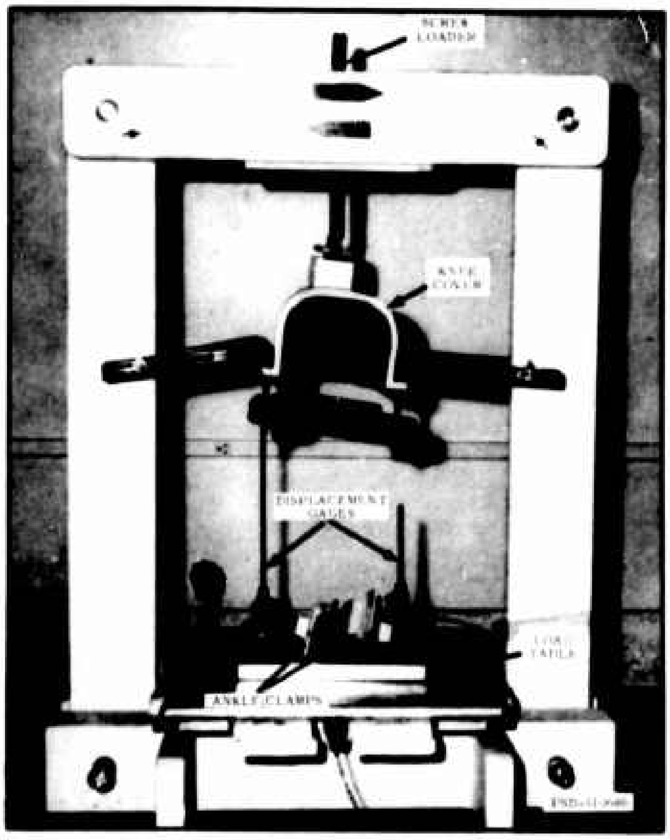

Figure 4 – Compression Stand for Foot Compressibility Tests

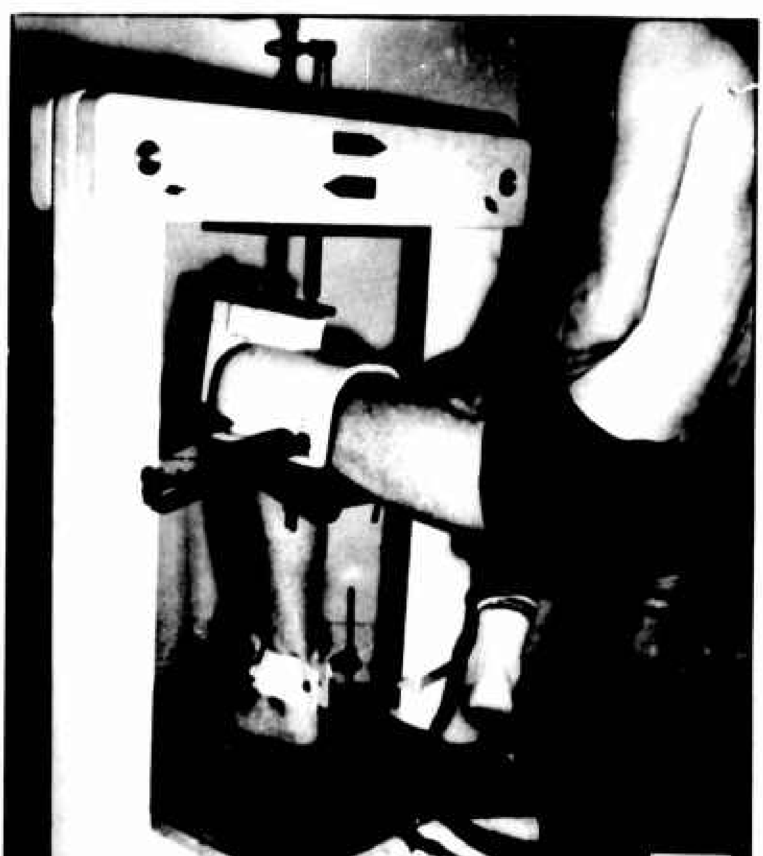

Figure 5 – Experimental Setup for Foot Compressibility Tests

Figure 6 – Footprint of Subject 5 for Three Knee Loads

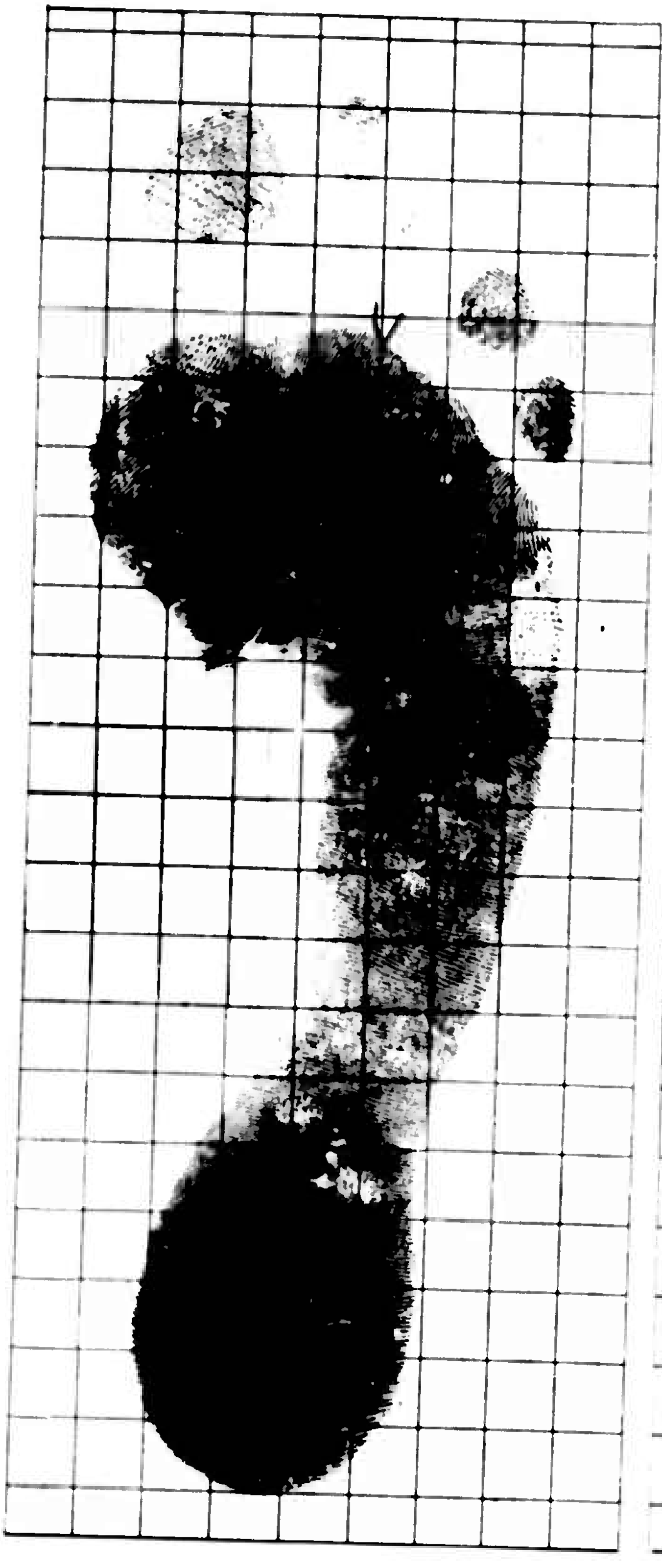

Figure 6a – Knee Load of 92 Pounds (One-Half of Body Weight), Area = 15.4 Square Inches

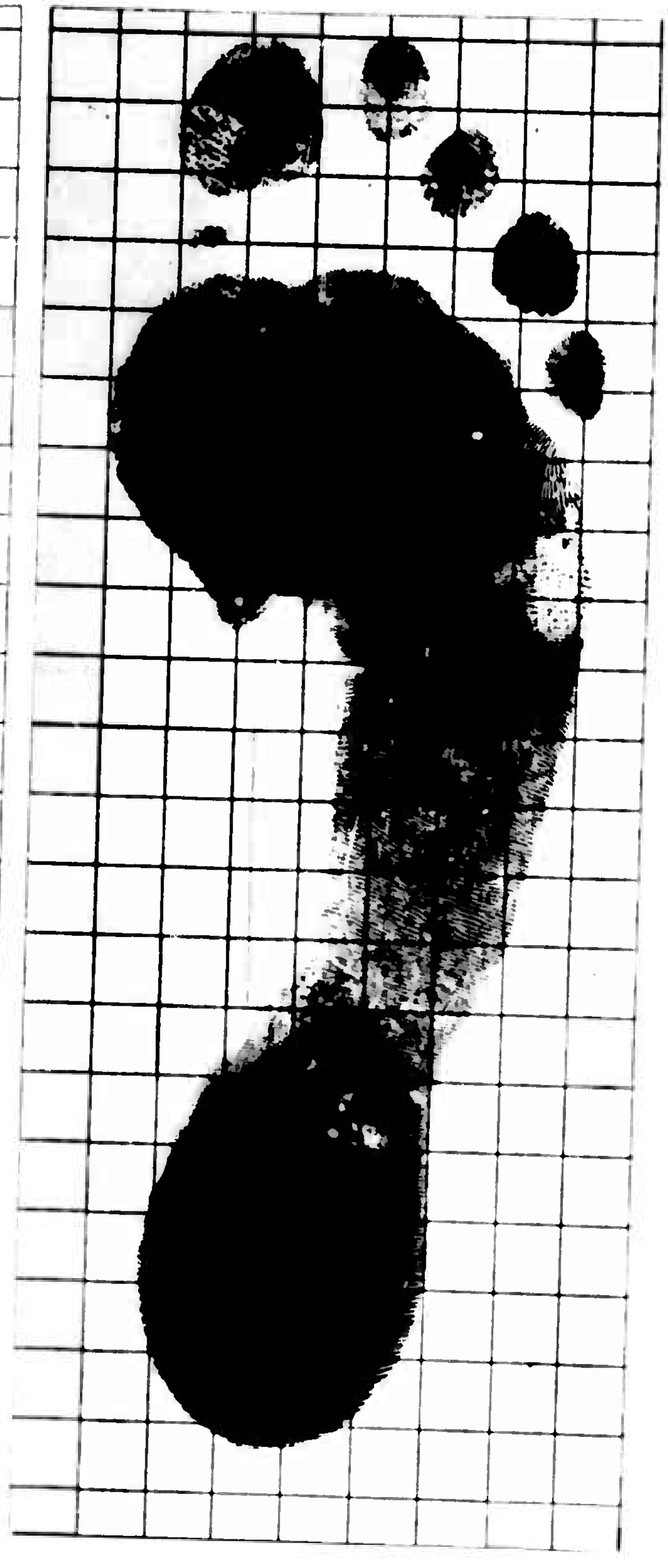

Figure 6b – Knee Load of 185 Pounds, Area = 17.7 Square Inches

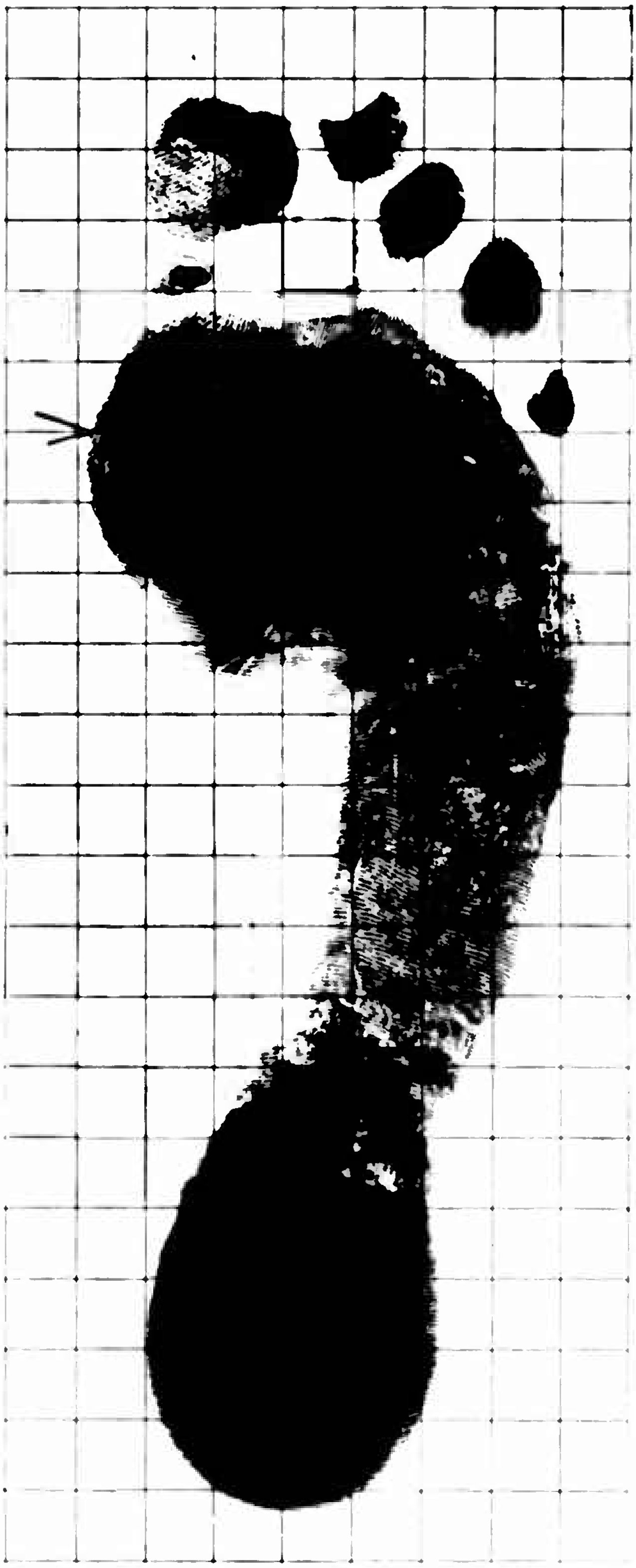

Figure 6c — Knee Loa.' of 340 Pounds,
Area = 18.3 Square Inches

RESULTS AND DISCUSSION

At the start, it may be useful to describe the construction of the skeletal system under study here (see Figure 1). Included are the iliac crests which are at the top and sides of the pelvic girdle. At the bottom of each side of the pelvic girdle, there is a roughly hemispherical hollow into which the head of the thigh bone (femur) fits. Below the femur are the tibia (shin) and fibula which, in turn, rest on the ankle and foot bones.[3] The vertical deflection of the iliac crest is a measure of the total compression of the structure below, and the slope of a plot of the applied loads versus the measured deflections would be a measure of the overall spring constant (stiffness K) of the lower limbs.

The major elements of the supporting structure of the lower limbs can be considered a series of articulated rod-like bones with softer cartilagenous material lining the bearing surfaces at the articulations. It is apparent that in compression, such a structure will act as a series of springs for each leg.

A highly simplified mechanical analogue is proposed in Figure 7. Here each leg component is envisioned as an elastic structure: the bones including the femural neck are very stiff (in loading tests on a whole human femur 40 cm long, Marique[4] found that the compressive shortening of the inner border will not exceed 1 mm even with loads large enough to cause fracture). The cartilagenous structures are less stiff, and the fleshy heel pad is a very soft easily compressible cushion. A further simplification is made, i.e., that connective tissues can be assumed to deform elastically in some approximately linear fashion so that

$$F = Kx$$

where F is the applied load in pounds and x is the change of length in inches. It will be apparent from the measurements which will be introduced that the human material under consideration does not follow such simple laws. It is felt however that for carefully limited ranges, such approximations are adequate to yield values which can be useful inputs for the computations of dynamic response of man.

First let us consider Test Condition 1 in which the stiffness of the total system shown in Figure 7 (i.e., both lower limbs) was measured. The vertical deflections of the hip as a function of shoulder loads in excess of normal body weight are shown in Figure 8. Although the individual data point plots show some scatter, there is a clear indication that for at least two subjects, a linear approximation of the curve slope is not a bad match. For Subject 3 there is considerable deflection with small load until about 75 lb above body weight; after this point, the slope becomes comparable to the others. As will be shown in subsequent sections, this is probably due to an individual variation in heel pad structure. The heel pads of Subjects 1 and 2 have reached the end of their compressibility (bottomed) with body weight, while the heels of subject 3 can continue to compress for an additional 75 lb above body weight.

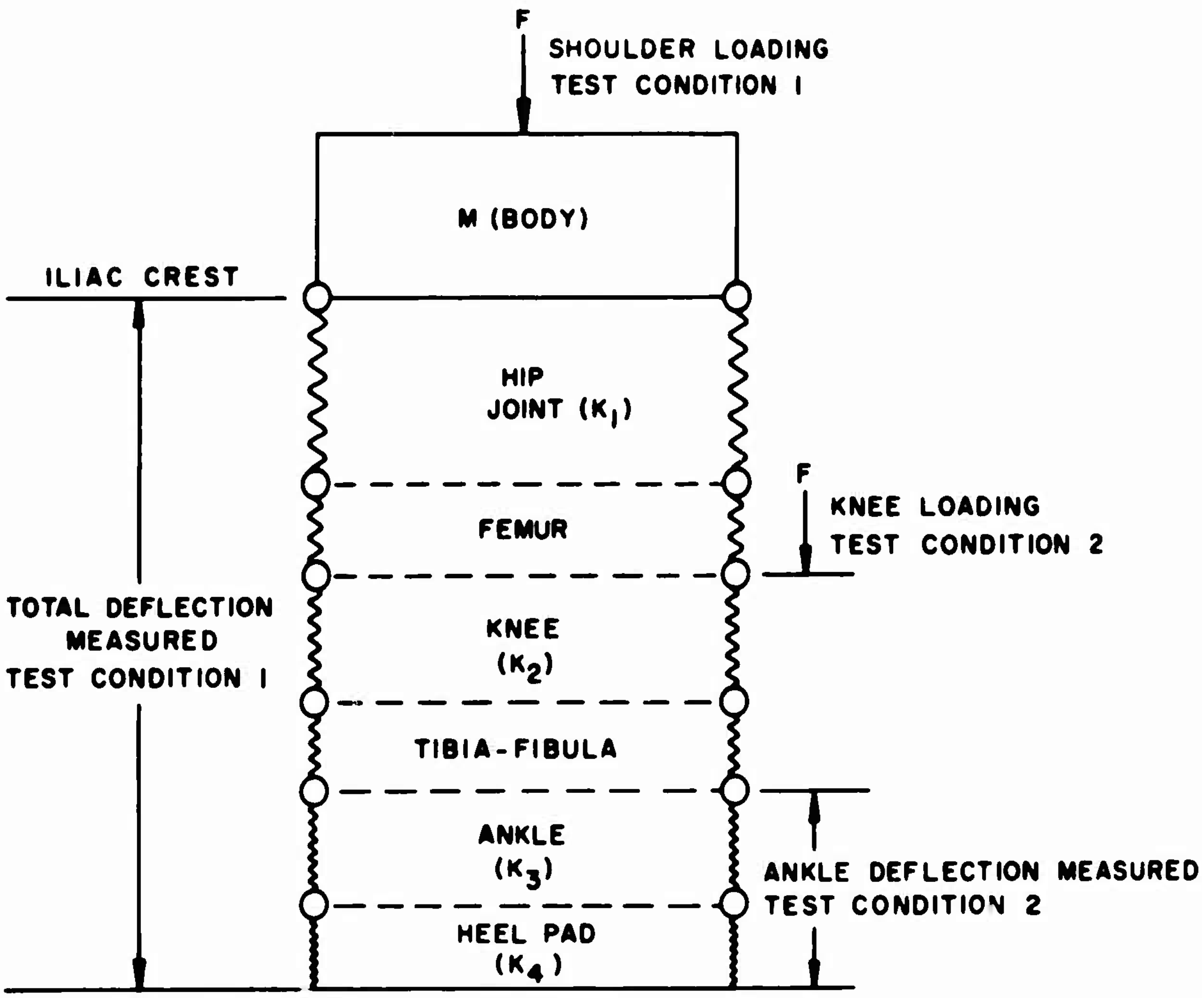

Figure 7 – Simplified Mechanical Model of Lower Limb Structure

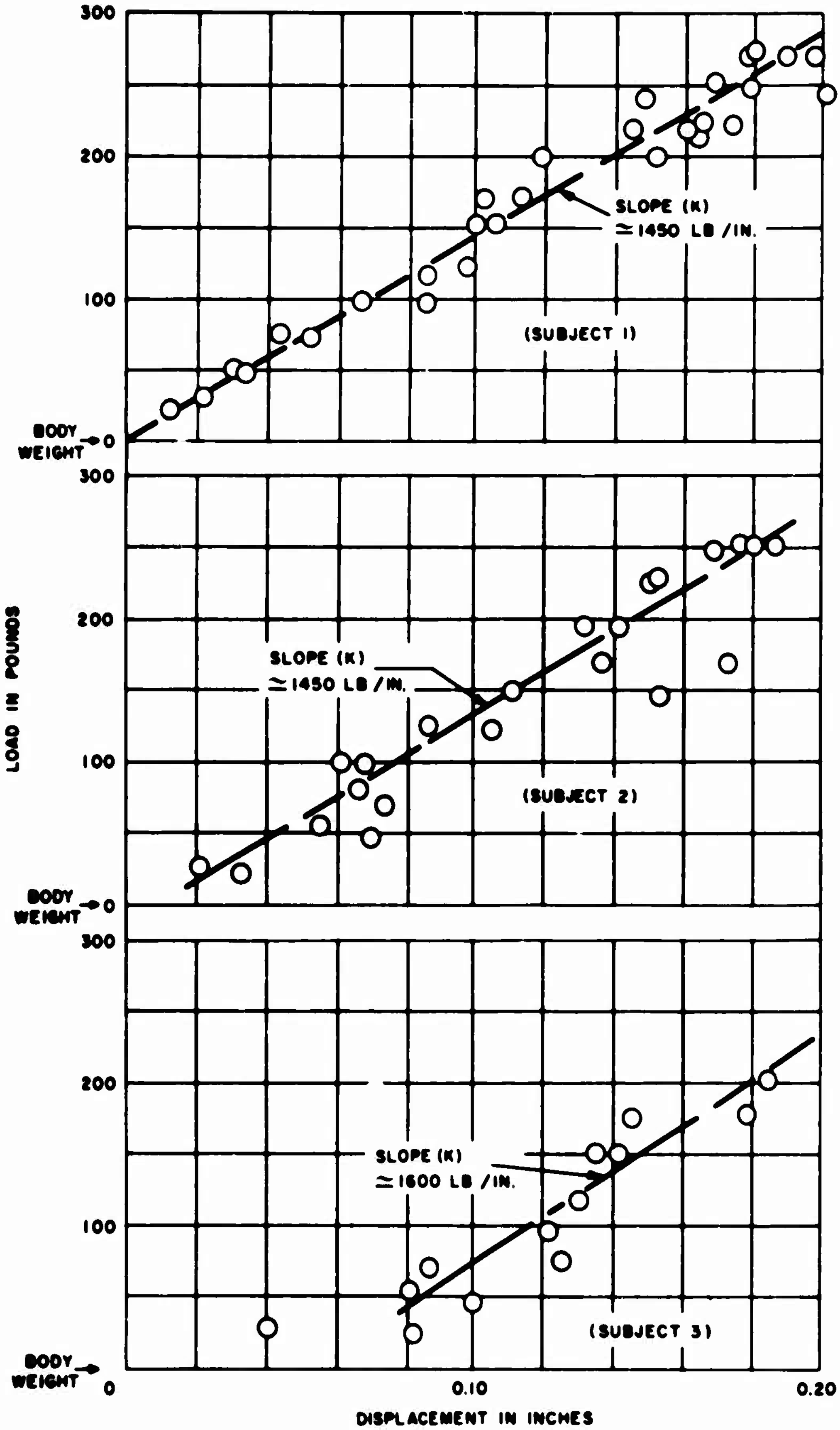

Figure 8 – Deflection of the Iliac Crest under Vertical Load

(Note: Initial conditions correspond to the load and displacement due to body weight.)

The described data indicate that in the range of total lower limb loading between the normal body weight and 200 to 280 lb above body weight, a linear curve can be drawn which will describe compressive response, and a single average value of 1500 lb/in. for total stiffness K_T of the lower limbs can be assigned. This value of stiffness is sufficient for dynamic calculations of total body response as described in the introduction.

It was of interest to determine in which region the bulk of the leg deformation was taking place, i.e., below the ankle, the hip, the knee, or distributed among these joints. The second set of tests, already mentioned, provided data on the compression of the foot and ankle joint. A composite curve including the displacement of the ankle with load for eleven test runs on five subjects is shown in Figure 9. The expanded curves in Figure 10 for loads up to about 100 lb, approximately the normal body loading on one foot, show that considerable ankle deflection occurs.

It should be noted that the loading range in the foot test which can be compared directly with the shoulder load test lies between approximately one-half the body weight and maximum load per leg, or one-half the total shoulder load. The curves for this equivalent loading condition are shown in Figure 9 as Region 2, lying between about 100 lb and 200 to 250 lb for the knee load.

An expanded plot of Region 2 is shown in Figure 11. The curve is nonlinear at lower values of load and there is a considerable spread of slopes. A reasonable average curve has been faired-in which shows a slope of about 2300 lb/in. between extremes of about 1700 to 4000 lb/in.

Measurements of total limb stiffness K_T (previously described) gave an average value of about 1500 lb/in. for two limbs in parallel. The stiffness of one limb K_L is half as great or 750 lb/in.

We can see from Figure 7 that one limb is composed of a series of at least four springs K_1, K_2, K_3, and K_4 neglecting the compliance of the bones. If we assume that in the range under consideration, K_4 (the heel pad) has to a large extent bottomed or become infinitely stiff, then we are left with the usual equation for three springs in series:

$$1/K_L = 1/K_1 + 1/K_2 + 1/K_3 \qquad [1]$$

where K_1 is the stiffness of the cartilage at the hip joint,

K_2 is the stiffness of the knee cartilage, and

K_3 is the ankle stiffness.

To facilitate the calculation, K_1 and K_2 can be lumped into one value A, i.e., $\frac{1}{A} = \frac{1}{K_1} + \frac{1}{K_2}$. Rewriting Equation [1] as

$$1/K_L = \frac{1}{A} + 1/K_3$$

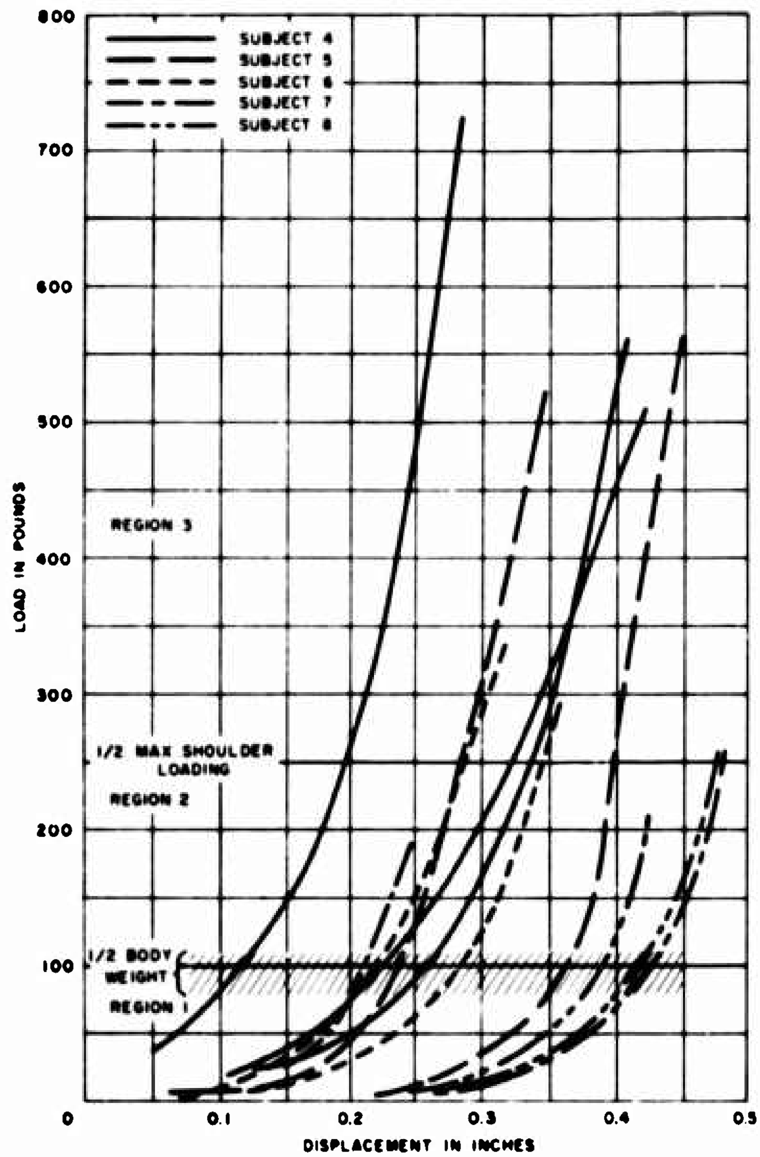

Figure 9 – Composite Plot of 11 Tests of Deflections of the Ankle under Vertical Load

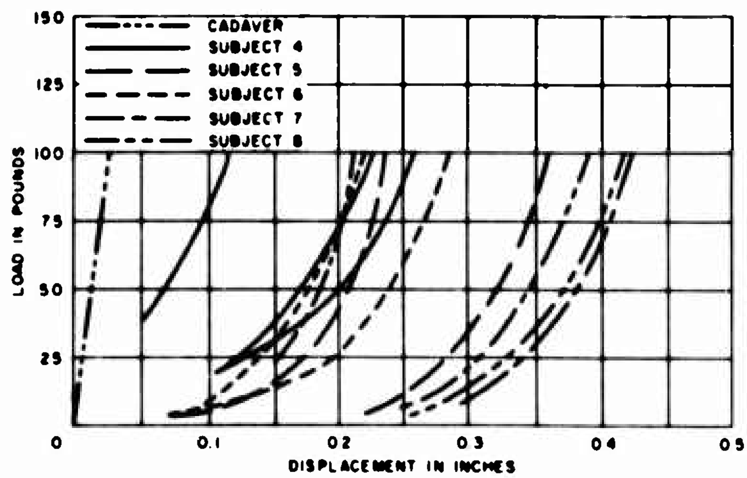

Figure 10 – Composite Plot of 11 Tests of Ankle Deflection under Compressive Loading at the Knee from Zero to Equivalent Body Weight of 100 Pounds

The cadaver data as taken from Reference 5.

Body weight assumed to be about 200 pounds.

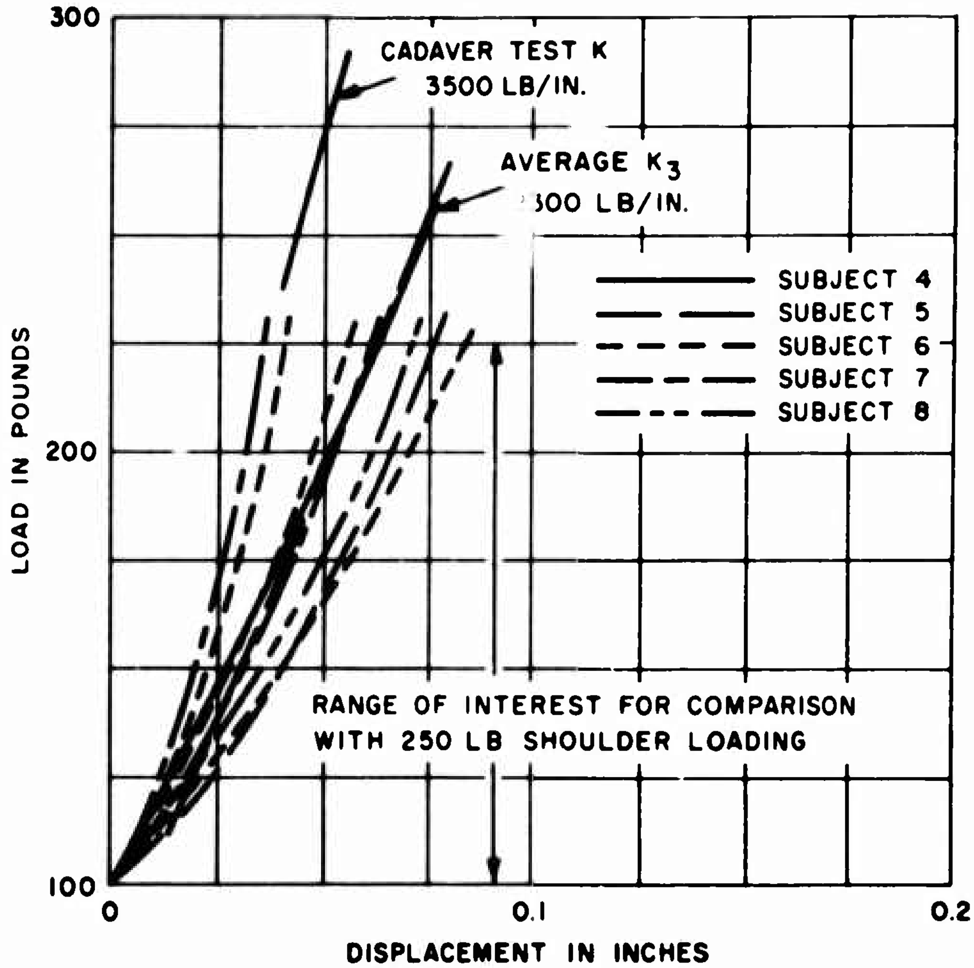

Figure 11 – Ankle Displacements Adjusted to Common Origin at Equivalent Body Weight of 100 Pounds

Subjects under compressive loading at knee of 100 to 275 pound (Region 2).

we can substitute the values for K_L of 750 lb/in. and K_3 (ankle stiffness) of 2300 lb/in. and find that $\frac{1}{A} = 1/1100$. The actual thicknesses of the hip and knee cartilages are about the same (5 to 7 mm or 0.20 to 0.28 in.); hence, we may expect the compressibility of the knee and hip to be similar and K_1 equal to K_2. Then $\frac{1}{A} = 1/1100$ and $K_1 = K_2 = 2200$ lb/in.

As a final matter of interest in this study, data from a cadaver test[5] similar to the knee-loading tests described here are presented in Figure 12. These measurements were made during a compression test on one leg which was brought up to about 1500-lb load at which point a fracture occurred in the distal tibia. The slope or stiffness in the loading region of interest in this study appears to be about 3500 lb/in. The data are included in Figures 10 and 11 for comparison purposes. It can be seen that the cadaver foot is significantly stiffer in both regions.

Since the study reported in Reference 5 took place over a long period of time, it is assumed that the cadavers were preserved rather than fresh although this point is not specified.

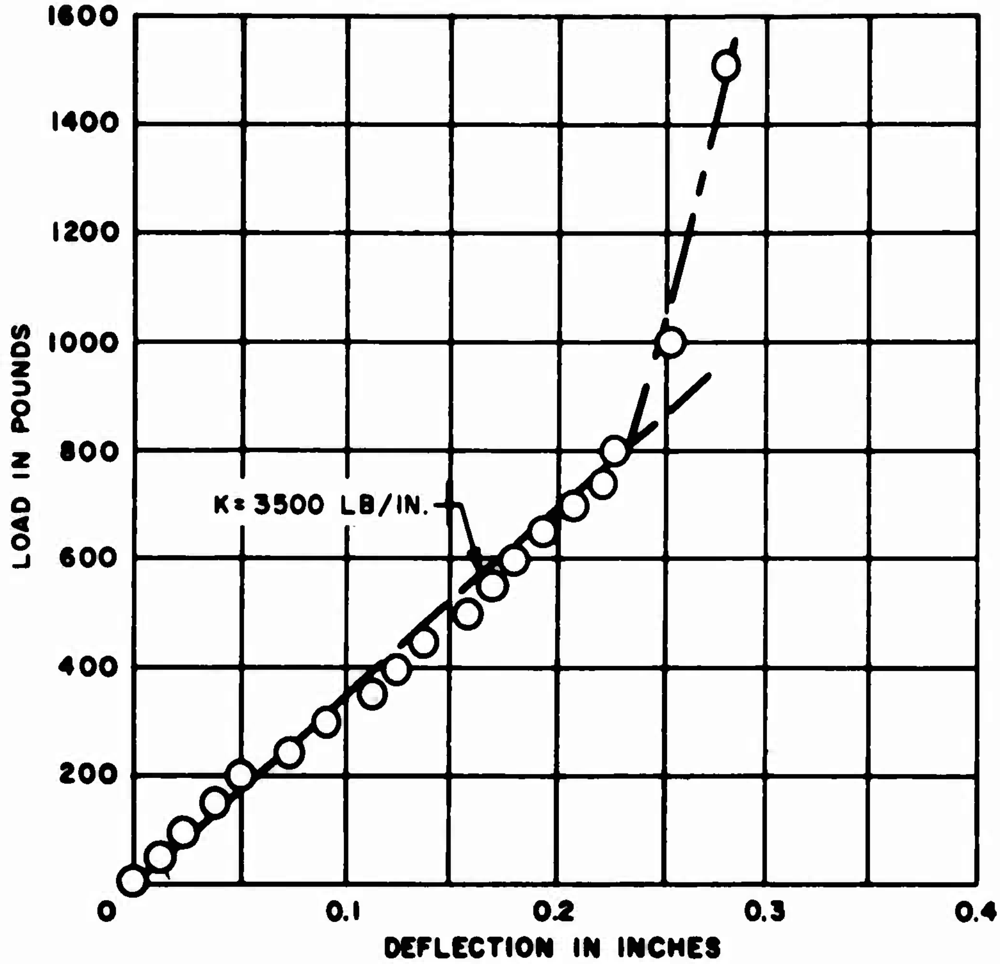

Figure 12 – Static Load Deflection Curve of Ankle on Cadaver Leg

(After Draeger, Reference 5.)

Consequently it can be expected that the normal joint fluids would be replaced by preservative and that joint cartilages would be shrunken and dessicated, retaining only slight elasticity. Hence we would not expect to find the heel pad compression of Region 1 that was present in volunteer tests. This expectation is borne out, as can be seen from the curve of cadaver heel compression shown in Figure 10. The cadaver ankle also exhibits a greater stiffness as can be seen in Figure 11.

It has been shown that no more than about one-third of the total limb compression occurs below the ankle. Is this deformation, as has been assumed, taken up by the cartilagenous material in the ankle-heel region or is it in bending of the longitudinal arch? Some researchers[6, 7] have indicated that the arch is extremely strong and does not act as a spring. Although not conclusive, the data of Figure 6 seem to indicate only slight arch deflection. The prints of a foot under various conditions of load show that increasing a load by as much as 370 percent increased the footprint area by only 20 percent with a negligible change in the open area under the longitudinal arch. Most of the area increase occurred in the heel region with some change at the transverse arch. Flattening was most marked under the heel as evidenced by the considerable darkening of the heel area, indicating that the heel takes most of the load.

As further evidence of the major load-bearing function of the heel, the limiting loads were almost always determined by the onset of pain under the heel. This was most likely due to the pressure of the heel bone (calcaneal tuberosity) against the heel pad tissue. This phenomenon is due to the particular axis along which the load was applied in conditions of testing. A similar loading was applied in the cadaver test, where in static loading, failure occurred as a fracture of the distal tibia rather than in the arch ligaments. It should be pointed out that about 75 percent of the body weight during normal relaxed stance rests on the heels.[8] It is reasonable to expect that the major loads from shock motions will be transmitted into the body through this region.

This is substantiated by war damage studies[9] in which direct shock injury for standing man occurred either as fracture of the calcaneum, distal tibia or fibia rather than injury to bones or ligaments of the longitudinal arch.

Some data were obtained on hysteresis and on loading rate. Figure 13 shows results of tests on Subject 4. In the case of this man, there seems to be some indication of greater initial stiffness with a faster application of load; however, measurements with other subjects show much scatter, and no definite estimate of the effect of loading rate can be made.

CONCLUSIONS

1. When supporting weights up to 280 lb in excess of normal body weight, the two lower limbs are approximately equivalent to a linear spring with a constant K of about 1500 lb/in., i.e., each leg has an equivalent spring constant of 750 lb/in.

2. When subjected to loads in excess of body weight equivalent, the average spring constant for one foot and ankle is about 2300 lb/in.

3. When the lower limbs are loaded with weights up to about 280 lb above body weight, foot compression will account for about one-third of the total vertical deflection of the hip bone relative to the floor.

ACKNOWLEDGMENTS

The authors wish to acknowledge the assistance of Mr. R. Durkovic, formerly of the Model Basin, who set up and recorded data for the leg compressibility test, Miss B. Johnson, Mr. R. Mahone and Mr. F. Whitten who provided engineering support, Drs. D. Drum and R.B. Gresham, USNMC, and D.E. Goldman, CAPT (MSC) USN, who rendered technical assistance.

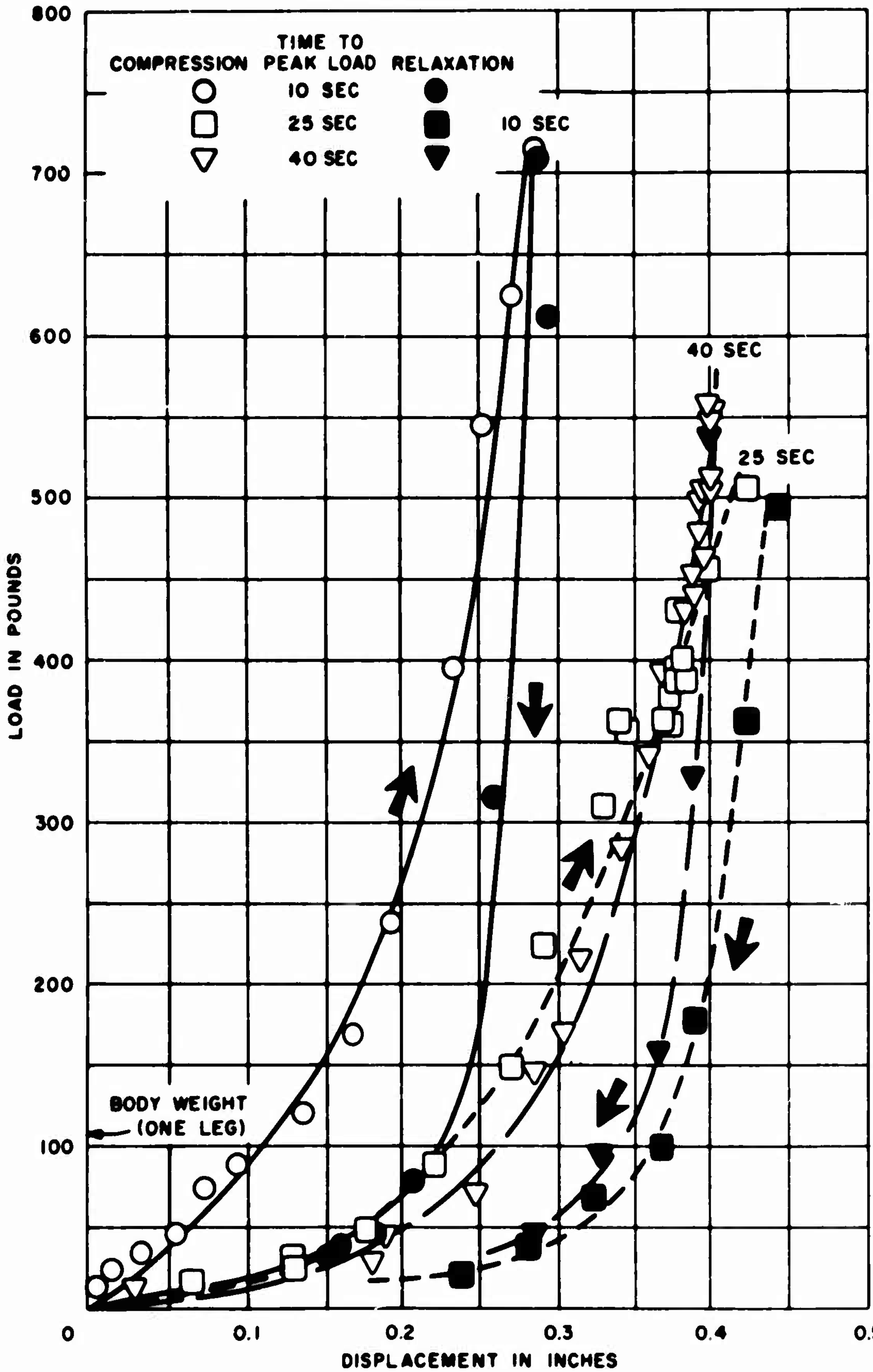

Figure 13 – Deflection of Ankle (Subject 4) under Different Loading Rates

REFERENCES

1. Coermann, R.R., "The Mechanical Impedance of Human Body in Sitting and Standing Position at Low Frequencies," Aerospace Medical Laboratory, Wright-Patterson, AFB, Ohio Project 7231, Task No. 71786, ASD Technical Report No. 61-492 (Sep 1961).

2. Hirsch, A.E., "The Effect of Shock Motions on Man," Paper No. 63-WA-283, Philadelphia, Penna., Human Factors Society of the ASME (Nov 1963).

3. Goldman, D.E., and Von Gierke, H.E., "The Effects of Shock and Vibration on Man," Naval Medical Research Institute, Lecture and Review Series No. 60-3 (Jan 1960).

4. Marique, P., "Etudes sur Le Femur," Librairie Des Sciences Bruxelles, Rue Coudenberg 76–78 (1945).

5. Draeger, R.H., et al., "A Study of Personnel Injury by Solid Blast and the Design and Evaluation of Protective Devices," Naval Medical Research Institute, Research Project X-517, Report No. 1 (Mar 1945).

6. Gratz, C.M., "Tensile Strength and Elasticity Test of Human Fascia Lata," Journal of Bone Joint Surgery 13: 334–340, (1931).

7. Lapidus, P.W., "Kinesiology and Mechanical Anatomy of the Tarsal Joints," Clinical Orthopaedics and Related Research, No. 30, J.B. Lippincott Company, Philadelphia, Pa. (1936).

8. Hicks, J.H., "The Three Weight-Bearing Mechanisms of the Foot," Chapter 7, Biomechanical Studies of the Musculo-Skeletal System," F. Gaynor Evans, Editor, Charles C. Thomas, Publisher, Springfield, Illinois (1961).

9. Hirsch, A.E., and Johnson, B., "Personnel Casualties Resulting from Underwater Explosive Attacks Against Naval Surface Ships (U)," David Taylor Model Basin Report C-1582 (Jul 1963) CONFIDENTIAL.

INITIAL DISTRIBUTION

Copies

8 CHBUSHIPS
- 3 Ship Protection (Code 423)
- 1 Lab Mgt (Code 320)
- 2 Tech Lib (Code 210L)
- 1 Stru, Ship Prot, Hull Mats & Fab (Code 341A)
- 1 Pers Prot and Mine Sweep (Code 622A)

2 CNO
- 1 OP-323T
- 1 OP-323D

2 CHONR
- 1 Bio Sci (Code 440)
- 1 Physiology (Code 441)

1 CHBUMED
- 1 Code 75

1 CO & DIR, USNRDL
- 1 Code 911

1 CO & DIR, USNMDL

2 CO, USNMRI
- 2 D.E. Goldman, CAPT (MSC) USN

9 CHDASA
- 1 FCTG
- 3 DASAMD
- 3 DASAABS
- 1 Document Lib

20 DDC

2 FAA, Oklahoma City, Oklahoma
Attn: Dr. J. Swearingen

2 NASA
- 1 Dr. E.B. Konecci
- 1 Code RBH
  Mr. G.N. Chatham

2 Wright-Patterson Air Force Base Ohio
Aerospace Medical Laboratory
Attn: Dr. H. von Gierke

Copies

2 USN ACEL
Philadelphia, Penna.
Attn: Dr. Hendler

1 Biomechanics Research Center
Wayne State University, Detroit
Attn: Dr. H.R. Lissner

1 Lovelace Clinic, 4800 Gibson Blvd
S.E. Albuquerque, N.M.
Attn: Dr. C. White